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A 12-station, anatomic hip joint simulator

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ABSTRACT

A novel 12-station hip joint simulator with an anatomic position of the prosthesis was designed and built. The motion of the simulator consists of flexion-extension and abduction-adduction. The load is of the double-peak type. The validation test was done with three similar 28 mm CoCr/polyethylene joints in diluted calf serum lubricant for 3.3 million cycles. The bearing surfaces of the polyethylene cups were burnished, the CoCr heads were undamaged, the wear particles were in the 0.1–1 μm size range, and the mean wear factor of the polyethylene cups was 5.7×10^{-7} mm^3/N m. These essential observations were in good agreement with clinical findings. In addition, three similar 50 mm CoCr/CoCr joints, representing the contemporary large-diameter metal-on-metal articulation were tested. The wear of the CoCr/CoCr joints was calculated from the Co and Cr concentrations of the used lubricant quantified with atomic absorption spectroscopy. The bearing surfaces of the CoCr/CoCr joints showed mild criss-cross scratching only. The average wear factor of polyethylene cups was 275 times that of the CoCr/CoCr joints. The tribological behaviour of the large-dia. CoCr/CoCr appeared to be dominated by fluid film lubrication, as indicated by very low frictional heating and wear, making it tribologically superior to the conventional CoCr/PE, and therefore very interesting clinically. In conclusion, the simulator proved to be a valid, reliable, practical, economical and easy to operate tool for wear studies of various hip replacement designs.

Keywords: hip joint simulator, validation, ultra-high molecular weight polyethylene, CoCr, large-diameter metal-on-metal articulation

INTRODUCTION

For laboratory wear studies on prosthetic hip joints, a number of different test devices called hip joint simulators have been built. Ten contemporary hip joint simulators were analysed and compared in [**1**–**4**]. In the present study, the basic idea was to design and build a multi-station hip joint simulator that is valid, reliable, practical, economical, and easy to operate. This machine design enterprise was motivated by the need of increasing the testing capacity of the author's laboratory. In the present paper, the outcome of the design effort, the HUT-4 simulator, is described in detail. The HUT-4 simulator is based on the valuable experience gained from its predecessors, HUT-1 [**5**], HUT-2 [**6**], HUT-3 [**7**], and from the HUT versions of the biaxial rocking motion simulator [**8**,**9**], originally introduced by McKellop and Clarke [**10**]. In the 12-station HUT-4 simulator, the prosthesis is in the anatomic position and is self-centring, the motion is electro-mechanical consisting of flexion-extension and abduction-adduction of the femoral component, the load is pneumatic, dynamic, vertical, fixed relative to the cup and continually measured, the lubricant chamber is large, and the fixation and removal of the test specimens is quick and convenient.

It is important that a new simulator is first shown to produce wear similar to that occurring clinically. For this purpose, it is rational to use materials and prosthetic designs, the clinical behaviour of which is known through several scientific publications. The HUT-4 simulator was validated with the most commonly used bearing couple, 28 mm CoCr against conventional ultra-high molecular weight polyethylene. Moreover, a new prosthetic design representing the contemporary large-diameter metal-on-metal articulation that has aroused a broad interest among orthopaedic surgeons, was included in the tests in order to evaluate its tribological behaviour in comparison with the conventional CoCr/PE.

MATERIALS AND METHODS

With the 12-station HUT-4 simulator (Fig. 1), the tribological behaviour of any type and size of prosthetic hip joint can readily be studied. The femoral head holder is fixed to an inner, abduction-adduction (AA) cradle (Fig. 2). The AA axis is perpendicular to that of the outer, flexion-extension (FE) cradle. There are four interconnected FE cradles, each having three AA cradles. The 12 AA cradles are linked together. The excursions of the FE and AA motions are 46° and 12°, respectively, in accordance with biomechanical studies of walking [**11**–**13**]. The motions are implemented with a specially developed crank mechanism driven by an electric motor via a gear. Both motions are nearly sinusoidal, and their phase difference is $\pi/2$ (Fig. 3). The motions were measured with calibrated angle transducers while the simulator was running normally. The cycle frequency is 1.06 Hz. Because FE is made by the outer cradle, and AA by the inner cradle, the Euler sequence of the motions is FE→AA, according to the definition of orthopaedic angles [**1**,**2**]. The internal-external rotation (IER) has earlier been shown to be unimportant in the simulation of clinical wear [**3**,**4**,**9**], and so it was left out. The crucial requirement is that the direction of sliding changes continually, and two motion components, even with the present simplified waveforms, are sufficient for that. The head is easily and accurately aligned with the FE and AA axes with a positioning disk (Fig. 2, lower left corner) that can be moved in three orthogonal directions. After the alignment has done with the help of a measuring gauge, the positioning disk is firmly fixed to the AA cradle. The head holder is fixed to the positioning disk, that accurately matches a recess on the bottom of the holder, by a single screw. Hence, the holder together with the head is extremely easy to remove for periodic cleaning, examination and wear measurement, and to reinstall in exactly the same position for the continuation of the test. It was checked by removing and reinstalling the holder several times that the centre of the head remained at the intersection of the FE and AA axes to an accuracy of 0.01 mm. The inclination of the neck axis of the femoral head holder is 45º, in accordance with typical femoral stems.

The acetabular cup is stationary and located above the femoral head in an anatomic position with respect to inclination and anteversion angles (Fig. 2). The cup is loaded with a pneumatic cylinder via a loading bar and a universal joint that makes the cup self-centring on the head. Hence, a misalignment of the head even much larger than 0.01 mm would not affect the test results. The universal joint is fixed to the slider of a vertical linear guide. The slider is pushed downwards by the cylinder. The direction of load is vertical and fixed relative to the cup. The load is dynamic, of double-peak type, and is measured continuously with a force transducer (Fig. 3), calibrated for compression using a specially made double frame and a 5 kN weight. The force signal is monitored with a digital oscilloscope. Only one test station is equipped with a force transducer. This is sufficient because (a) all 12 cylinders are similar, (b) they are all controlled by a single heavy-duty proportional pressure regulator, (c) the tube length from the manifold to each cylinder is uniform, (d) the output of each cylinder was checked in situ with the force transducer, and (e) the type of the prosthesis was not found to affect the measured load. The control signal for the pressure servo loop of the regulator is generated by a cam fixed to the drive shaft of the crank mechanism, a roller follower, a displacement transducer and its amplifier. The cam shape was tailored to result in a double-peak load waveform similar to that measured with instrumented hip prostheses for normal walking [**14**], the maximum, minimum, and average load values being 2.0 kN, 0.4 kN, and 1.2 kN, respectively. In conformity with biomechanical studies [**11**–**14**], the load starts to rise at maximum flexion, the temporal distance between the two load peaks is 1/3 of the cycle time, the peak value is 2.5 times the body weight of 800 N, the second peak coincides with maximum extension, and the minimum value is clearly above zero. With the present motion and load, the so-called force track, i.e., the theoretical track of the resultant force vector on the femoral head is elliptical (Fig. 4), having an aspect ratio of 3.8, which has earlier been shown to represent human gait well [**3**,**8**]. The length of the force track is 1.73r, where r is the head radius. Considering the force track, the value of the integral $\int L\mathrm{d}x$, where L is the instantaneous load and x is the relative motion, over one cycle is is 2.09r N m (r to be substituted in mm). The present motion is similar to that of the BRM simulator [**3**,**4**], with the exeption that the AA amplitude is 12º instead of 46º. This makes the force track shape an ellipse instead of a circle [**2**,**3**].

The acrylic lubricant chamber is open, and its volume is large, 500 ml. These are important characteristics in order to avoid the overheating and excessive degradation of the serum-based lubricant, which could lead to erroneous wear mechanisms [**15**,**16]**. The bottom of the chamber is sealed to the head holder with a silicone o-ring. A plastic cover above the chamber reduces the risk of contamination.

The lubricant was prepared so that triple 0.1 μm sterile filtered, non-iron supplemented HyClone Alpha Calf Fraction serum, catalog no. SH30212.03, lot no. ANA18095, was diluted 1:1 with Milli-Q-grade distilled water. Hence, the protein concentration of the lubricant was 21 g/l. No additives were used. The Fe concentration of the serum was 0.45 mg/l. Each chamber contained 500 ml of lubricant. The lubricant temperature T_L during the test was not controlled, but T_L and the environment temperature T_E were regularly measured. T_L was measured c. 1 cm below the surface, and T_E c. 5 cm above the lubricant chamber. The quantity of interest was $\Delta T = T_L - T_E$.

Two types of tests were done, 28 mm CoCr against conventional ultra-high molecular weight polyethylene, and 50 mm CoCr against CoCr (Table 1). All specimens were manufactured and supplied by Sulzer/Centerpulse/Zimmer. The CoCr/PE test was a validation test for the new simulator. The 28 mm CoCr heads were polished. The polyethylene cups were machined from a compression-moulded sheet, were metal-backed, 12 mm thick, and gamma-irradiated by 25 – 40 kGy in nitrogen. Initially, their internal diameter was 0.1 mm larger than the head diameter. The cups were similar to those used in an earlier study [**9**].

The 50 mm CoCr cups were from the Durom resurfacing system, whereas the 50 mm CoCr heads were designed for the use with a femoral stem, as an alternative to a Durom resurfacing femoral component, the bearing surfaces of the two heads being identical. The bearing surfaces of the 50 mm CoCr heads and cups were polished. For the calculation of clearance, the diameter of the heads was measured with a micrometer, and the internal radius of the cups was measured with a coordinate measuring machine. The head diameters were 49.95–49.97 mm, whereas twice the internal radius of all three cups was 50.13 mm. Hence, the diametral clearances of the couples varied from 160 μm to 180 μm, and relative clearance from 3.2 ‰ to 3.6 ‰. The initial deviations from roundness of the heads and cups were around 1 μm, measured with a Talyrond 31C roundness measurement device. With deviations of this magnitude, the measurement uncertainty was of the order of 0.1 μm.

Both types of heads were fixed to the head holders by taper-fit. A piece of acrylic bone cement was cast on the outer surface of the titanium shells and of the CoCr cups in a mould so that a flat loading surface was formed for the flat end of the vertical loading bar (Fig. 2). The horizontal locking of this load interface was implemented with two guide pins. Hence, the acetabular component was even easier to remove and reassemble than the femoral head holder. The shells and the CoCr cups were positioned in the mould before the hardening of the bone cement so that in the test, their inclination angle was 45º and anteversion 15º, which are typical values clinically.

Three similar joints of both types were tested for 3.3 million cycles. The test was stopped at intervals of 550 000 cycles (six days) for cleaning, visual and microscopic examination, and wear measurement. After that, the test was continued with fresh lubricant. The wear of the polyethylene cups was measured using a gravimetric method published elsewhere [**17**]. The combined wear of each CoCr/CoCr couple was evaluated by measuring the Co and Cr concentrations of the used serum lubricant by atomic absorption spectroscopy (AAS, Varian 220 Zeeman, Centre for Chemical Analysis, Helsinki University of Technology). The AAS method was identical to that used in an earlier study [**18**]. The detection limit for both Co and Cr was 0.01 mg/l. Each concentration value recorded was the mean from three samples. To obtain an estimate of the total amount of material removed from the bearing surfaces, the sum of the Co and Cr concentrations was multiplied by the measured total volume of the lubricant. This value was multiplied by 1.1, as Co and Cr constitute 90 weight % of the Metasul alloy. All wear products were assumed to be in the lubricant.

In the CoCr/CoCr tests, the femoral head holder and the loading bar, both made of stainless steel 329 containing 25 % Cr (being the only metallic components apart from the test specimens that were exposed to the lubricant), were diamond-coated to inhibit metal ion release from them. To make sure that the measured Co and Cr originated from wear, and not from passive metal ion release from the Metasul components, head holder, or loading bar, a control test was done so that all conditions were similar to the wear tests, with the exception that the joint was not loaded, and there was no relative motion between the head and the cup. Moreover, the Co and Cr concentrations of the fresh serum were quantified with AAS.

With both types of joints, the wear rate was taken to be the slope of the linear regression line in the diagram wear vs. number of cycles, the first 550 000 cycle stage being omitted in the regression as a transient phase (running-in). The wear factor was calculated so that the wear rate, in mg per one million cycles, was divided by the density of the material, the integral $\int L\mathrm{d}x$, and 10^6. The densities of Sulene-PE and of Metasul alloy were 0.94 mg/mm^3 and 8.29 mg/mm^3, respectively. The values of the integrals $\int L\mathrm{d}x$ over one cycle for the 28 and 50 mm joints were 29.3 N m and 52.3 N m, respectively.

Polyethylene wear particles were isolated from samples of used lubricant by NaOH digestion, HCl neutralization, and filtration on 0.05 μm pore size Nucleopore filters, photographed with a scanning electron microscope, and studied with image analysis tools as described elsewhere [**19**].

RESULTS

In visual examination and in optical microscopy, the contact zone of the polyethylene cups appeared polished (Fig. 5). The 28 mm CoCr heads were undamaged. They were not scratched, and did not show any patches or layers. Due to the large lubricant volume, the polyethylene wear particle concentration in the used lubricant was so low that in the micrographs of polycarbonate filters, only few particles were visible. Fig. 6 is a typical image showing one agglomerate, one strip, and several globular particles of different sizes. The mean ECD of the wear particles was 0.49 μm (range 0.20–0.99 μm, n = 103) (Fig. 7). On the heads and cups of CoCr/CoCr joints, only mild criss-cross scratching and some serum residue were observed on the contact zones (Fig. 8).

The wear of the CoCr/CoCr joints proved to be so low that it could not be reliably quantified by Talyrond measurements.

The AAS showed that the in the fresh serum, the Co and Cr concentrations were below 0.01 mg/l. In the used lubricant of the first 550 000 cycle stage of the CoCr/CoCr wear test, the average Co and Cr concentrations of were 5.65 mg/l and 2.53 mg/l, respectively. In the used lubricant of the succeeding five stages, the average Co and Cr concentrations were much smaller, 0.62 mg/l and 0.31 mg/l, respectively. The average Co/Cr ratio in the used lubricant was 2.1, equal to the Co/Cr ratio in the Metasul alloy indicating that Co and Cr detected by AAS really originated from wear. In the control test without load and relative motion, the Co concentration of the lubricant was below 0.01 mg/l, indicating that the passive metal ion release from the Metasul components was negligible. However, the Cr concentration was slightly increased, being 0.018 mg/l, probably because the diamond coating did not completely inhibit the metal ion release from the head holder and loading bar both made from stainless steel.

The wear rate of both types of joints after the first 550 000 cycle stage was fairly constant (Fig. 9). The average wear rate of polyethylene cups was 15.5 mg per one million cycles, whereas the average combined wear rate of the CoCr/CoCr couples was 0.89 mg per one million cycles. The wear rate ratio was therefore 17. The corresponding wear factors were 5.7 $\times 10^{-7}$ mm^3/N m and 2.1 $\times 10^{-9}$ mm^3/N m, the ratio being 275.

The mean value of $T_E \pm$ SD was 23.8 ºC ± 1.0 ºC. The mean value of $\Delta T \pm$ SD in the CoCr/PE and CoCr/CoCr tests were 3.1 ºC ± 0.6 ºC and 1.3 ºC ± 1.5 ºC, respectively. According to the Wilcoxon rank-sum test, the difference was statistically significant. Surprisingly, ΔT in the CoCr/CoCr test was sometimes negative, the observed minimum value being –1 ºC, as the energy removed by evaporation outweighed that generated by friction.

DISCUSSION

The novel 12-station hip joint simulator HUT-4 was validated with the clinically most commonly used material couple, CoCr against conventional ultra-high molecular weight polyethylene. The bearing surfaces of the polyethylene cups were burnished, the CoCr heads did not show tarnishing, patches, layers or any other kind of damage, the wear particles were in the 0.1–1 µm size range, and the mean wear factor of the polyethylene cups was 5.7 $\times 10^{-7}$ mm^3/N m. These essential observations agreed with clinical findings [**20**–**22**]. The size distribution of the particles (Fig. 7) was highly similar to that of particles recovered from periprosthetic tissues [**20**, Fig. 13]. The mean wear factor in retrieved Charnley polyethylene cups has been found to be 2.1 $\times 10^{-6}$ mm^3/N m [**22**], substantially higher than the mean value in the present study. The difference is likely to be attributable to two factors, (1) the stainless steel femoral heads of the Charnley prostheses were often scratched by acrylic particles leading to the abrasive wear mechanism and to increased wear rate, whereas the CoCr heads of the present tests remained undamaged, the adhesive wear being the dominating mechanism, and (2) the wear resistance of the polyethylene used in the early Charnley prostheses was probably not as good as that of the modern Sulene-PE used in the present tests. CoCr heads can also be scratched in vivo by third bodies [**20,23**], which, however, were absent in the present tests. The clear border between the inferior part of the polyethylene cup where the orginal machining grooves were still intact, and the superior part polished by wear (Fig. 5) was strikingly similar to that found in cups removed from patients [**20,24**]. This was naturally due to the anatomic position of the cup in the simulator, which resulted in a realistic location and size of the contact zone. A similar observation was made by Dowson and Jobbins with their three-station, three-axis Leeds hip joint simulator [**25**]. In both the Leeds and the HUT-4 simulators, the prosthesis is in the anatomic position, and the FE cradle contains three inner, AA cradles. In a cup worn in a BRM simulator, there is no such border, but the entire hemisphere is polished [**4**,**9**,**20**]. Therefore, the contact area in the HUT-4 simulator is smaller and maximum contact pressure higher compared with the BRM. Still these two simulators produced mean wear factors for similar materials very close to each other, 5.7×10^{-7} mm^3/N m (HUT-4) and 5.2×10^{-7} mm^3/N m (BRM) [**9**]. Hence, the elliptical force track with an aspect ratio of 3.8 (HUT-4) and the circular force track (BRM) resulted in fairly similar wear factors despite the differences in contact area, sliding speed and $\int L dx$. This is in good agreement with a recent pin-on-disk study, in which the effect of the slide track aspect ratio on wear factor was investigated [**26**].

The polished appearance of the bearing surfaces of the CoCr/CoCr joints with mild criss-cross scratching and patches of serum residue (Fig. 8) were in agreement with clinical retrieval studies on the old large-dia. McKee-Farrar CoCr/CoCr designs [**27**,**28**], that had much higher variation in sphericity and clearance compared with the present specimens. Those McKee-Farrar prostheses that by chance had an optimal clearance could work very well for 25 years or so, but sometimes the mismatch was so severe that early failure occurred. The measured deviations from roundness of the present 50 mm CoCr heads and cups, that were of the order of 1 μm, are tribologically insignificant in relation to the diametral clearances that varied from 160 to 180 μm. Moreover, the deviations did not increase during the tests. Apparently, now that the deviations have been reduced to the micrometre level with modern manufacturing technology, it is the clearance that determines the tribological behaviour. With optimal clearance, the hard-on-hard couple with sufficient diameter appears to work on fluid film lubrication, at least for some part of the cycle [**29**].

The wear products of CoCr/CoCr differ significantly from those of polyethylene. The CoCr particles are typically at least one order of magnitude smaller than the polyethylene particles, and so they are biologically less active [**30**,**31**]. Moreover, the CoCr particles are likely to dissolve with time and leave the system. The polyethylene particles, however, being chemically inert remain in the periprosthetic tissues and keep provoking a continuous macrophage reaction, which can lead to substantial ostelysis and loosening of the implant fixation necessitating a revision operation, the results of which are, unfortunately, most often poor. AAS does not distinguish particles from ions, but quantifies the total element concentration, which in fact is the very subject of interest. AAS has been used in the quantification of the Co and Cr concentrations in the blood of patients with low-dia. CoCr/CoCr prostheses. Increased concentrations compared with controls were found [**32**,**33**], but the systemic effects of these concentrations that are of the order of 1 μg/l are disputed.

CoCr/CoCr joints similar to the present ones were tested with an AMTI simulator [**34**]. In those tests, it was found that running-in wear of tens of micrometers occurred, after which the wear was negligible. In the present tests, there was no detectable increase in the deviations from roundness relative to the original values that were of the order of 1 μm. Hence, the AMTI simulator produced much more wear for these joints compared with the HUT-4 simulator. However, the differences in the composition of the serum-based lubricant may be even more important than the differences in the simulator designs. In the AMTI simulator tests, the lubricant was calf serum diluted 1:2 with buffered saline, whereas the present lubricant was alpha calf fraction diluted 1:1 with distilled water. In general, it can be stated that presently there is a need for much better understanding of the behaviour of different types of serum-based lubricants. No consensus exists among research groups regarding the preferred lubricant except for the vague requirement that it must contain proteins. This makes it very difficult to make meaningful inter-laboratory comparisons for wear results. Nevertheless, the observed difference between the wear rates of polyethylene and CoCr/CoCr was similar to that observed by another research group [**35**].

An international standard on hip joint simulators, ISO 14242-1 [**36**], has recently been published. The HUT-4 design deviates from this standard in several details, because it was found that many of the requirements of the standard have no scientific basis in the published literature of biotribology and orthopaedic biomechanics. First, the ISO standard requirement of the 1:3 dilution of the serum has been shown to result in unrealistic wear mechanisms, whereas the present 1:1 dilution ratio proved to be optimal [**37**]. Second, the encapsulation of the test joint recommended in the ISO standard would prevent the beneficial evaporation from the lubricant chamber. The evaporation removes frictional heat and reduces the risk of overheating of the contact surfaces and of the lubricant. This is why the HUT-4 lubricant chambers were left open. Third, the ISO standard requires that the cup abduction angle should be 30º instead of the clinically typical value 45º that was used also in the present tests, apparently in order to compensate for the vertical loading. It is often though that the loading direction should be inclined medially. However, the 30º cup abduction angle is not equivalent to the 15º inclination of the loading direction, because the former does not change the direction of the load relative to the motion axes [**2,3**]. Fourth, there is no evidence that the IER is necessary to produce clinically relevant wear, if the FE and AA are present and result in an elliptical force track (Fig. 4). The main point is that the direction of sliding changes continually, for which two motion components are sufficient [**2,3,8**,**9,19**]. For this reason, the IER was excluded, although it is included in the standard. With the 12-station AMTI simulator [**38**], it was shown similarly that it is rather the AA component that is particularly important in the reproduction of the clinical wear mechanisms and wear rates. The author's previous hip joint simulator design, HUT-3 [**7,8**], had the load inclined 12º medially, and three motion components, FE, AA and IER, but the comparison of the HUT-3 and HUT-4 results showed that the inclination and the IER are unimportant. Fifth, the system tests of the HUT-4 showed that the use of the ISO standard peak load of 3 kN resulted in a similar protuberance formation on the polyethylene bearing surface as the over-dilution of the serum [**37**], a phenomenon not seen in retrieved cups, but typical in wear tests done according to the standard [**39**]. The peak load was reduced to 2 kN, and as a consequence, the protuberances diaspepeared. Sixth, tests done at the ISO standard temperature of 37 ºC have never been shown to produce wear that would resemble clinical wear more closely than tests done at lower temperatures, such as the present ones, that are less prone to serum degradation problems, and can be run without toxic additives [**40**]. In summary, the development of wear simulation methods must continue, and hopefully will lead to an improved version of the ISO 14242 standard.

It was shown with the biaxial HUT-4 simulator having a simple, robust structure and consisting of simple, inexpensive parts and components, that expensive computer-controlled servo-hydraulics is not needed to produce valid wear simulation for prosthetic hips. With servo-hydraulics, it is easy to change the motion and load waveforms compared with the present way of implementing them. However, a fact too seldom stated is that there is no real need to change the motion and load waveforms, once a functional combination has been established. In practice, the simplicity brings excellent reliability and ease of operation. The present tests were most uneventful. There was not a single technical malfunction hampering the running of the test. The ease of operation had a high priority in the design.

In the following, the actions during a periodic wear measurement stop and during the running are described. After the test has been stopped, the top beam to which the loading cylinders are fixed is set aside, after which the slider of the vertical guide is pulled away together with the universal joint, loading bar, and acetabular cup. Then the head holder together with the head and the lubricant chamber is lifted away. The lubricant is poured to a glass container and then back to the chamber. This is repeated several times in order to rinse the surfaces exposed to the lubricant and to homogenize the lubricant for sampling. Then the lubricant volume is measured and samples taken. All parts that have been exposed to serum are washed up. The head and the cup can be examined with a microscope, and their roundness can be measured with a Talyrond apparatus using specially made holders. Polyethylene cup is vacuum desiccated and weighed using a standard procedure. The head holder is easily reassembled with no need for head re-alignment. After the lubricant has been poured into the chamber, the cup is immersed in a vertical position so that no air pocket will form inside it. Then the cup is turned over the head in its normal position, and the loading bar is lowered in its place. The loading beam is attached, and the test can continue. During the running, all that the operator needs to do is from time to time to measure the temperatures and add distilled water to compensate for evaporation. In the present tests, these were done once a day. Naturally, these could later be automatized, but since the trouble was so minimal, the automatization was considered unnecessary. The pneumatic loading system proved to be so stable that no adjustment of the load waveform (Fig. 3) was needed during running. The motion waveforms did not require monitoring because they were implemented electro-mechanically, and therefore were invariable.

CONCLUSIONS

A novel 12-station, anatomic hip joint simulator HUT-4 was introduced and shown to produce wear similar to that observed clinically with the most commonly used material combination, CoCr against conventional ultra-high molecular weight polyethylene. The simulator proved to be valid, reliable, practical, economical, and easy to operate. In the comparative test, the modern large-diameter CoCr/CoCr articulation proved to be tribologically superior to CoCr/PE. The tribological behaviour of the CoCr/CoCr articulation appeared to be dominated by fluid film lubrication, as indicated by very low frictional heating and wear. This is most promising because the tribological properties are known to be of primary importance for the clinical performance of prosthetic joints.

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APPENDIX

Notation

k	wear factor
r	radius of femoral head
x	relative motion
AA	abduction-adduction
AAS	atomic absorption spectroscopy
BRM	biaxial rocking motion
CoCr	cobalt chromium alloy
ECD	equivalent circle diameter
FE	flexion-extension
HUT	Helsinki University of Technology
IER	internal-external rotation
L	load
T_E	environment temperature
T_L	lubricant bulk temperature
ΔT	$T_L - T_E$
UHMWPE	ultra-high molecular weight polyethylene

CAPTIONS TO ILLUSTRATIONS

Fig. 1 12-station anatomic hip joint simulator ‘HUT-4’. Length of base beam is 1 620 mm.

Fig. 2 Station no. 2 of HUT-4 simulator with 50 mm CoCr/CoCr joint installed. Note anatomic position of prosthesis including 45º inclination of cup and of neck axis, and 15º anteversion of cup, large lubricant chamber, force transducer for continuous monitoring of load, universal joint making cup self-centring on head, and diamond coating of head holder and loading bar for inhibition of metal ion release.

Fig. 3 True, measured variation of flexion-extension (FE) and abduction-adduction (AA) angles, and load (L) with time/cycle time in HUT-4 hip joint simulator. Cycle time = 0.94 s.

Fig. 4 Flattened slide track patterns of HUT-4 simulator illustrating relative motion. Due to flattening, equator and rim dia. is πr instead of $2r$. Since track points are turned to plane about centre point of each figure, and since figures are relatively small compared with r, there is practically no distortion of shape and size of slide track figures in this flattening method (http://lib.hut.fi/Diss/2002/isbn9512260840/).

Fig. 5 Optical micrograph from border of contact zone of polyethylene cup after test. On left, original machining grooves. On right, surface polished by wear showing mild criss-cross scratching.

Fig. 6 Scanning electron micrograph of polyethylene wear particles on 0.05 μm pore size polycarbonate filter. Acceleration voltage 10 keV.

Fig. 7 Size distribution of polyethylene wear particles.

Fig. 8 Optical micrographs from centre of contact zone of 50 mm CoCr head.

Fig. 9 Variation of wear with number of cycles, (a) polyethylene cups, (b) CoCr/CoCr joints.

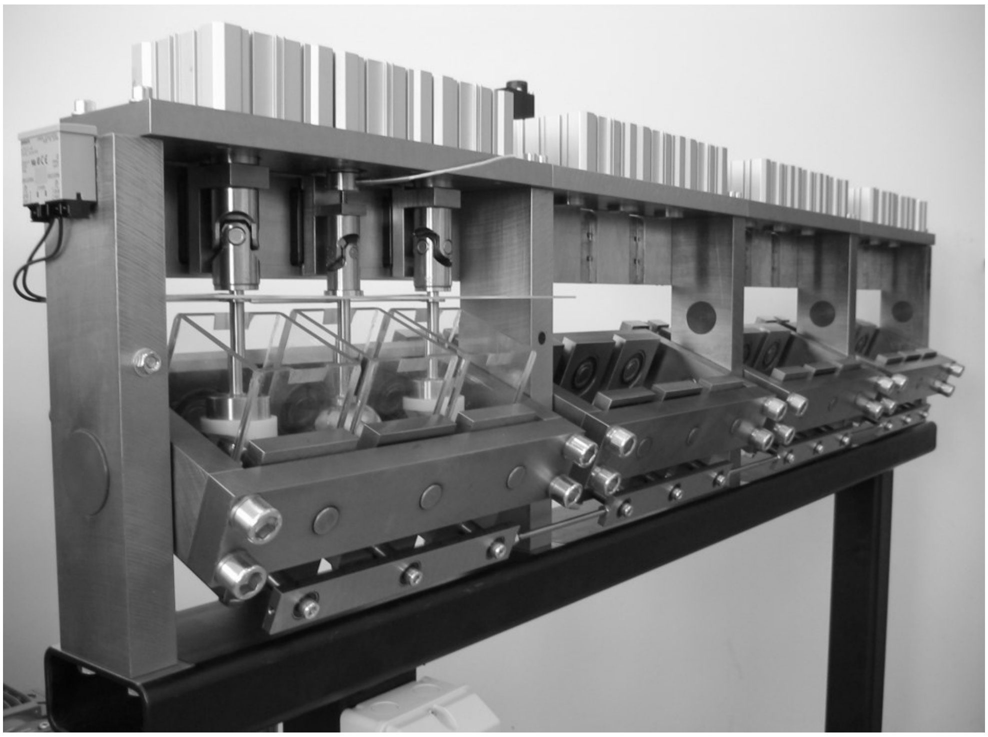


Fig. 1

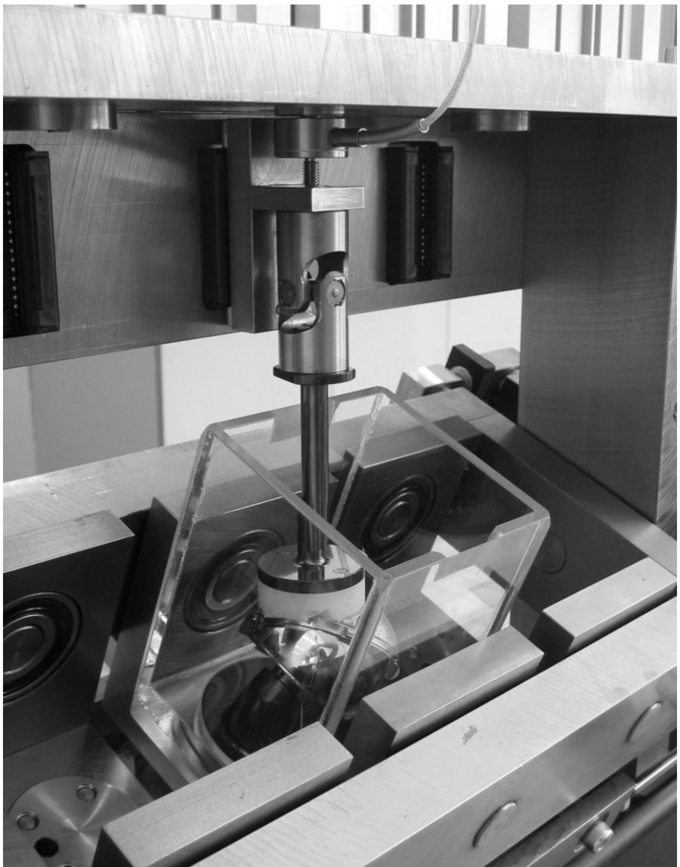


Fig. 2

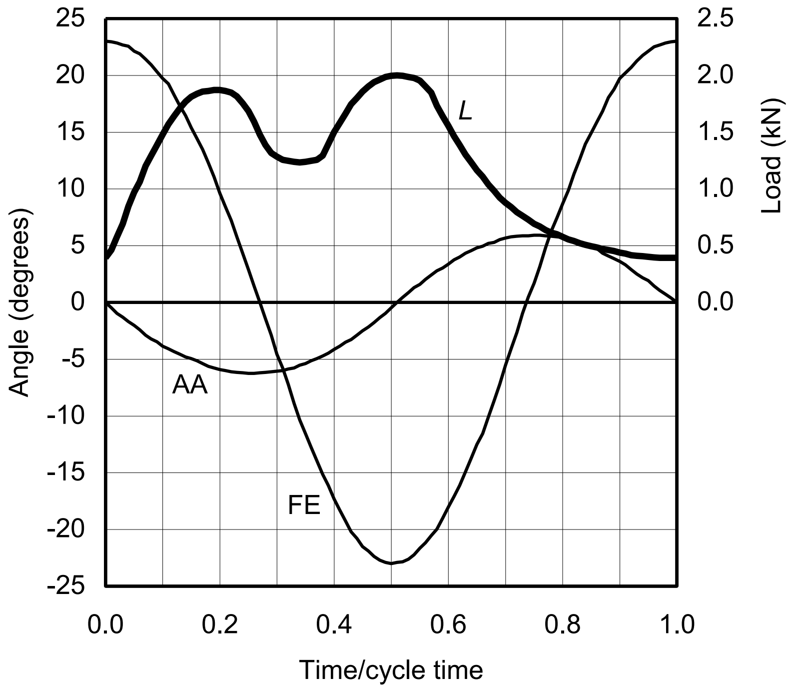

25
20
15
10
5
0
-5
-10
-15
-20
-25
2.5
2.0
1.5
1.0
0.5
0.0
0.0
0.2
0.4
0.6
0.8
1.0
Angle (degrees)
Load (kN)
Time/cycle time
L
AA
FE

Fig. 3

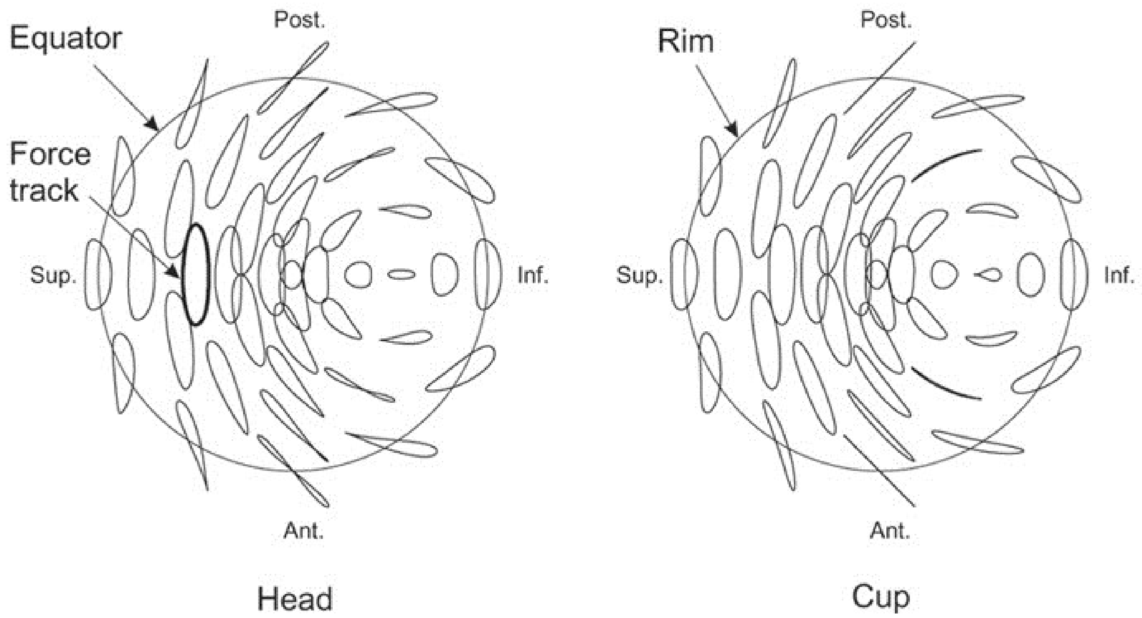

Equator
Force track
Post.
Sup.
Inf.
Ant.
Head
Rim
Post.
Sup.
Inf.
Ant.
Cup

Fig. 4

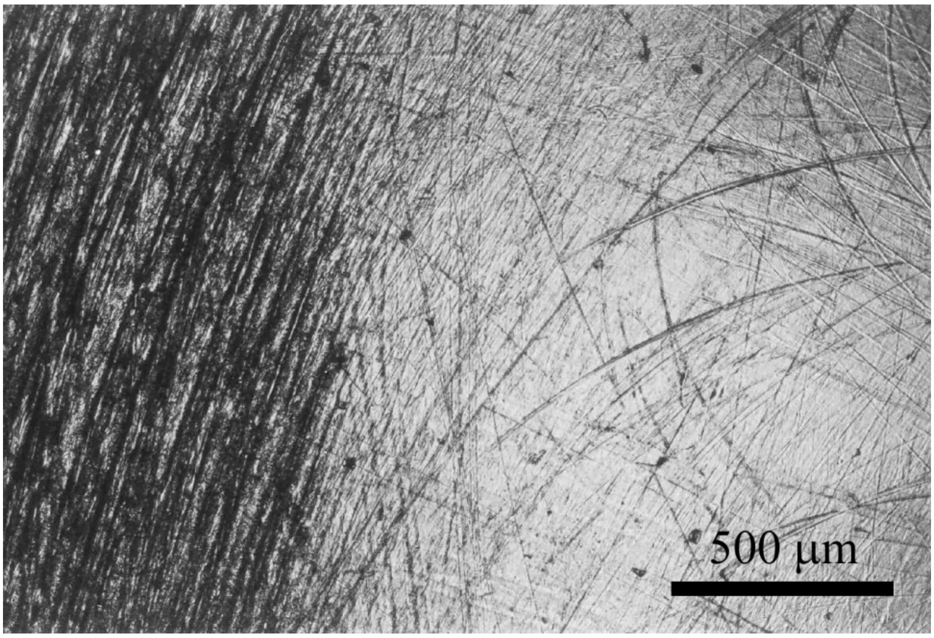


Fig. 5

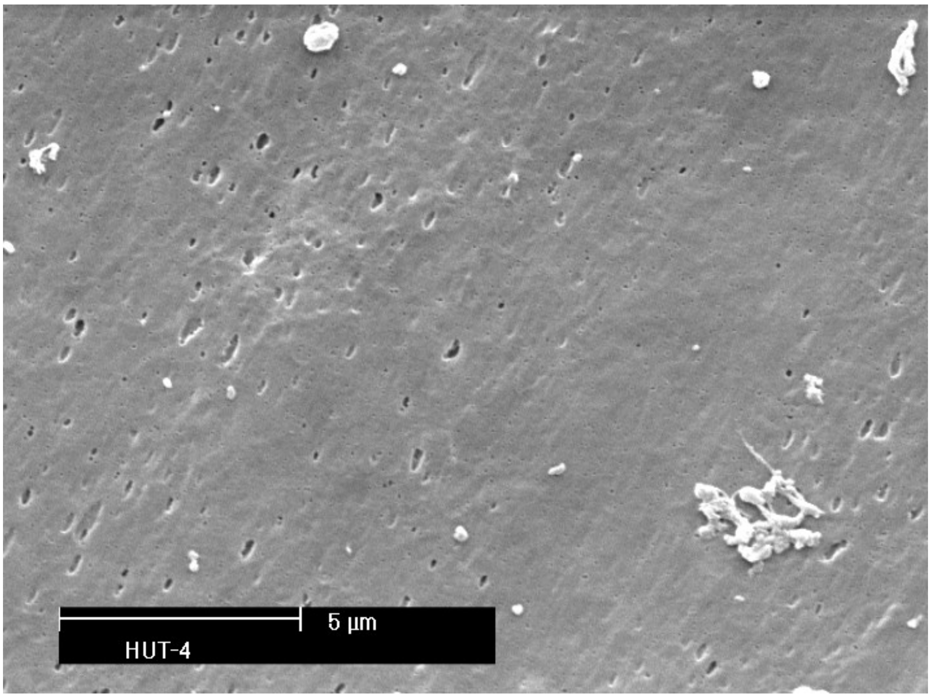


Fig. 6

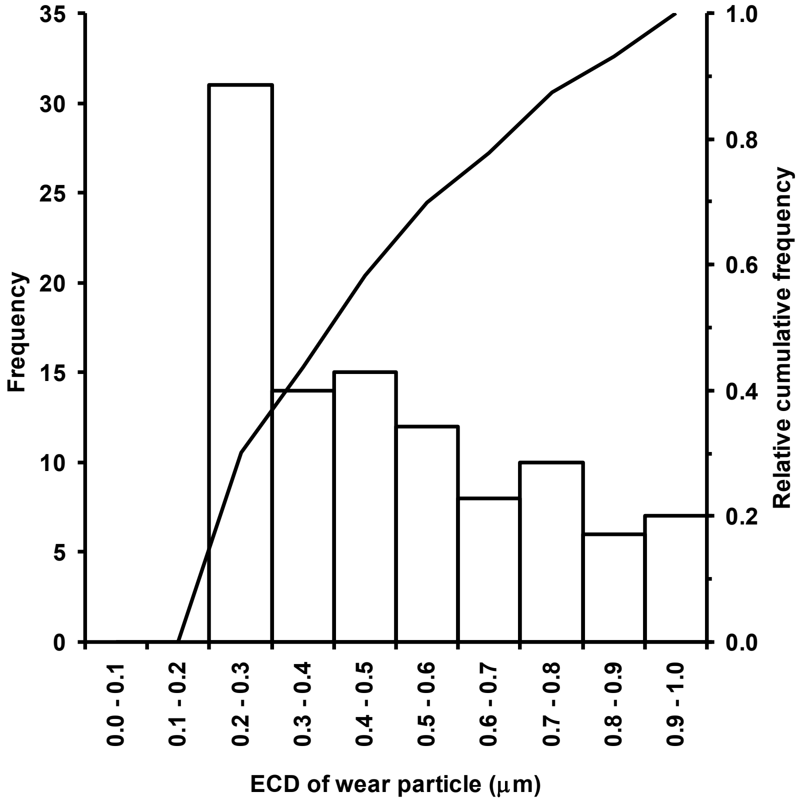

35
30
25
20
15
10
5
0
Frequency
1.0
0.8
0.6
0.4
0.2
0.0
Relative cumulative frequency
0.0 - 0.1
0.1 - 0.2
0.2 - 0.3
0.3 - 0.4
0.4 - 0.5
0.5 - 0.6
0.6 - 0.7
0.7 - 0.8
0.8 - 0.9
0.9 - 1.0
ECD of wear particle (μm)

Fig. 7

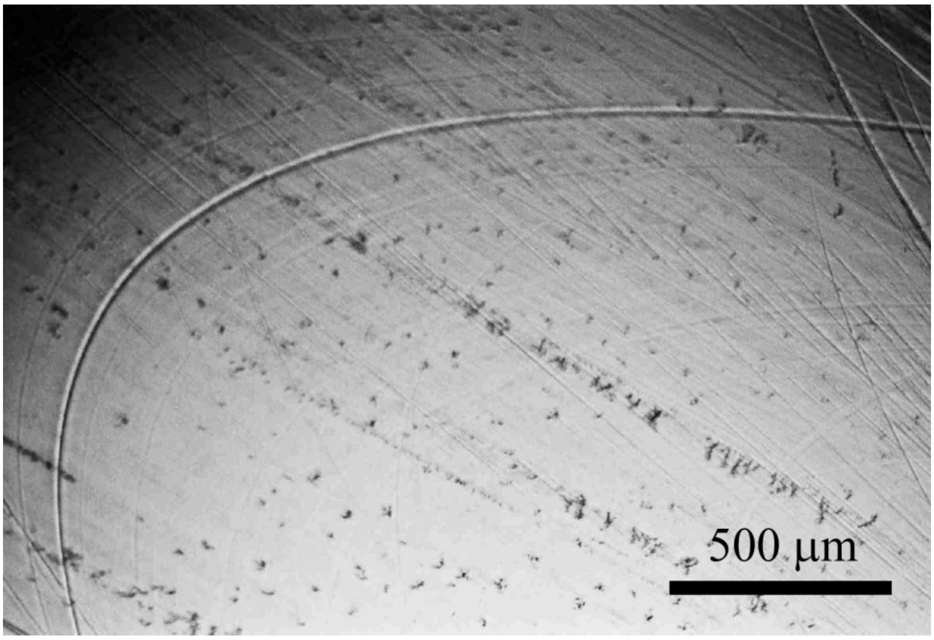


Fig. 8a

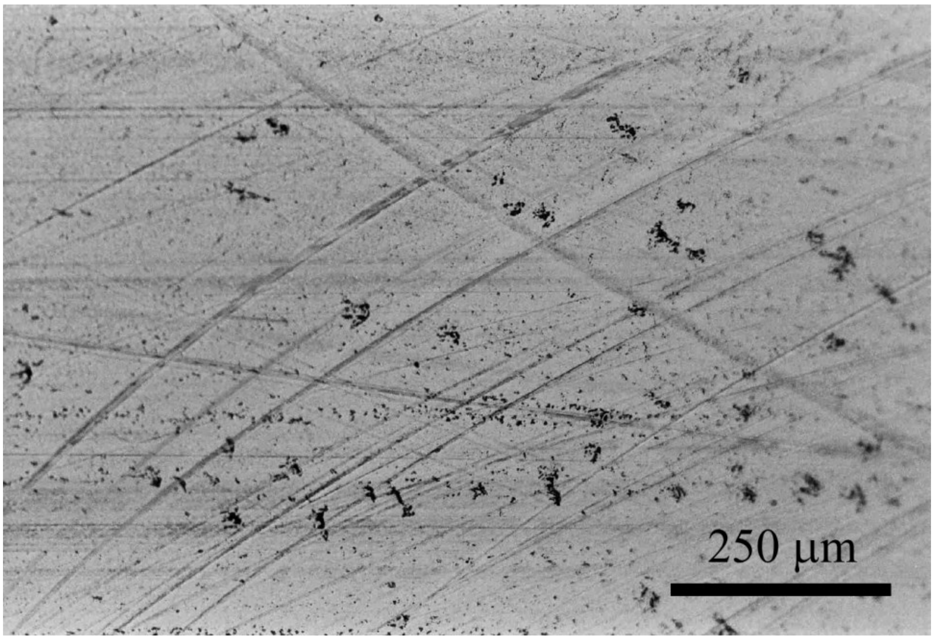
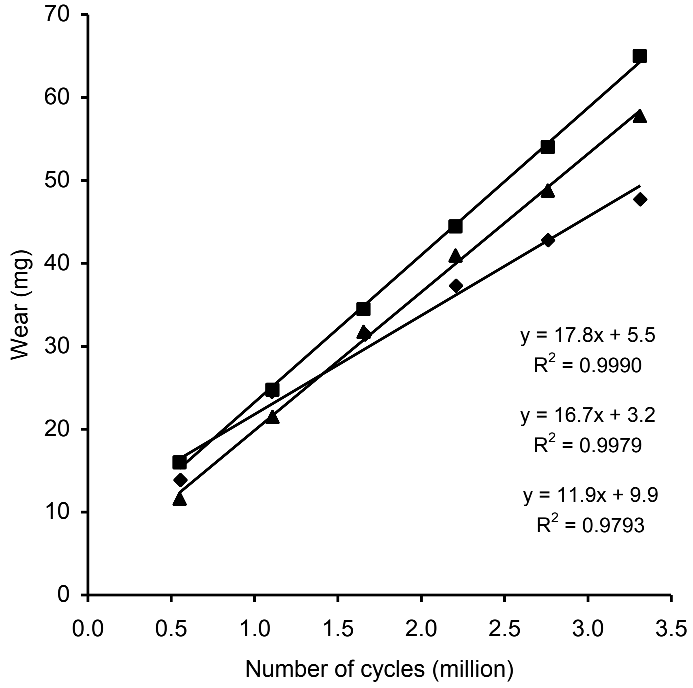
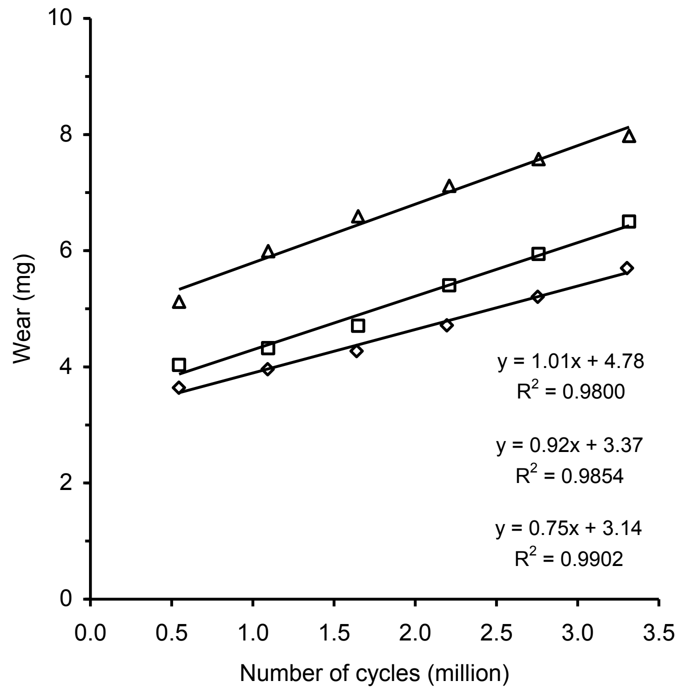


Fig. 8b



70
60
50
40
30
20
10
0
0.0
0.5
1.0
1.5
2.0
2.5
3.0
3.5
Wear (mg)
Number of cycles (million)
y = 17.8x + 5.5
R2 = 0.9990
y = 16.7x + 3.2
R2 = 0.9979
y = 11.9x + 9.9
R2 = 0.9793

Fig. 9a



10
8
6
4
2
0
0.0
0.5
1.0
1.5
2.0
2.5
3.0
3.5
Wear (mg)
Number of cycles (million)
y = 1.01x + 4.78
R2 = 0.9800
y = 0.92x + 3.37
R2 = 0.9854
y = 0.75x + 3.14
R2 = 0.9902

Fig. 9b

Table 1 Test specimens.

Type of joint	Component	Material	Mfr cat. no.
∅28 CoCr/PE	head	CoCrMo, hot-forged, low-carbon Protasul-20, ISO 5832-12	14.28.06-20
	cup	UHMWPE, GUR 1020, ‘Sulene-PE’, γ-sterilized in N_2, ISO 5834-1/-2	custom-made
∅50 CoCr/CoCr	head	CoCrMo, hot-forged, high-carbon Protasul-21WF ‘Metasul’, ISO 5832-12	01.00181.500
	cup	bearing surface material identical to that of the head	01.00214.056